Bicycle Helmet Efficacy Using Hybrid III and Magnesium Headforms

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ABSTRACT

Cycling is popular for transportation and leisure. Bike helmets are used to mitigate the risk of head injury in collisions or falls. According to NHTSA, 52,000 cyclists were injured in the US in 2010. Head injuries accounted for two thirds of hospital admissions, among cyclists. Despite strong epidemiological evidence of efficacy, some researchers and many lay people argue that bicycle helmets do not lower the head injury risk in bicycle crashes and falls. One definitive data set that would contribute to this debate is largely missing and that is biomechanical tests of helmeted vs. unhelmeted head impacts. This is because bicycle helmet standards specify magnesium headforms and unhelmeted impacts are not possible with these headforms. Therefore it was our objective to establish the Hybrid III headform as a valid device for evaluating bicycle helmet effectiveness by comparing it to a magnesium headform in impacts with contemporary bicycle helmets. The HIII head was also used in helmeted and unhelmeted tests to investigate overall bicycle helmet efficacy. A monorail drop tower was used to create drops from 0.5 m to 3.0 m in 0.5 m increments. Eight drops were performed using a helmeted magnesium headform and 16 were performed using a HIII headform (8 helmeted/8 unhelmeted). Linear head acceleration, Head Injury Criterion (HIC) and Injury Assessment Reference Values (IARV) for AIS 4+ brain injury were used to compare helmeted to unhelmeted head impacts. The magnesium headform peak head accelerations and HIC values were 5% and 9% higher than HIII respectively. The average of the peak head accelerations for helmeted impacts was 4.2 times lower with HIC values 6.8 times lower than the unhelmeted impacts. The HIII headform was found to reproduce the magnesium headform's time-domain response within an acceptable margin. Bicycle helmets were highly effective at mitigating and preventing injury in these tests.

INTRODUCTION

Cycling is a common mode of transportation and popular recreational activity with many known health benefits (Wen, 2008; Hamer 2008). Given its nature of moderate velocity travel over variable surfaces, cycling injuries are an inherent risk. In 2010, more than 52,000 cycling injuries were reported in the United States; 618 of those resulting in fatalities (NHTSA, 2010). In

Canada, cycling was determined to be the most injurious of all summer sports with 4,300 people hospitalised in 2010 (CIHI, 2010). Among cyclists in the US, head injuries account for approximately two thirds of reported cases and three quarters of fatalities (Thomspon, 2009).

Epidemiological investigations have shown clear support for helmets as a means of preventing head and brain injury amongst cyclists involved in falls or collisions. In a case-controlled study conducted by Thompson et. al., helmets were found to reduce head and brain injury risk by up to 85% and 88%, respectively (Thompson 1989). A larger study by the same group reported a reduction of head injury risk by 69% and brain injury risk by 65% when the cyclist wears a helmet (Thompson , 1996). Maimaris et. al. reported similar benefits in a study investigating over a thousand patients, with head injuries occurring in 11 percent of cases for non-helmeted cyclists compared to only 4 percent of the helmeted group (Maimaris, 1994). A Canadian study found that a person's risk of death due to a head injury incurred from a cycling accident increases three fold without a helmet (Persaud, 2012). In addition, helmets have been found to provide equal levels of protection in falls or collisions regardless of age group or motor vehicle involvement (Thompson, 2009).

Despite the support for helmet use through the epidemiological evidence, the safety benefits are not universally accepted. Many municipalities, provinces and states still do not have legislation for mandatory helmet use and some cyclists simply choose not to wear a head protection. One of the most prominent arguments for helmet efficacy is that certification standards only incorporate linear acceleration thresholds and neglect to address rotational acceleration, which some believe to be more relevant for brain injury (eg. diffuse axonal injury) (Ommaya, 1971; Adams, 1982). Curnow et. al. contends that epidemiological data does not distinguish between linear and rotational injuries (Curnow, 2003). Those against helmet efficacy also argue that the greater surface area of the helmeted head and the presence of vent holes lead to an increased risk of rotational injuries. Several researchers have published articles refuting the epidemiological investigations stating they are biased (Elvik, 2011; Curnow, 2006; Curnow, 2003, Hagel, 2003). Elvik criticises and evaluates a popular helmet use meta-analysis, accusing the authors of publication bias and stating the data had a time trend bias (Elvik, 2011).

Although injury statistics provide insight into cause and effect, they are limited to data collected retrospectively, after the impact event and thus are unable to control for the possible confounding variation in impact severity. Here we aim to biomechanically investigate the impact event itself to validate or refute the current epidemiological evidence starting from the premise that direct head impact will occur in the index cycling event.

In an effort to control the quality of helmets and design for bicycle-specific impacts, manufacturers adhere to certification standards set forth by various regulatory bodies. Although minor differences exist between the standards, they all use peak linear acceleration of the head as a biomechanical metric for impact severity. A helmet passes certification if the peak head acceleration remains below a threshold level when dropped from a prescribed height. The threshold value is not directly related to injury risk reference values and varies across the standards (CSA D113.2-M89, 2009; CPSC, 1998; ASTM F1447, 2006; Snell B95A, 1995; EN1078, 1997). The drop height is another variable parameter across these certification standards with the lowest being 1.5 m (EN1078) and the highest being 2.2 m (Snell B95A).

In several helmet efficacy studies, linear and rotational head accelerations are measured and effectiveness is determined by comparing metrics such as the Head Injury Criterion (HIC) to injury risk functions. Anthropomorphic test devices are common in these biomechanical investigations and are incorporated in many testing standards. The magnesium headform, used in many bicycle helmet certification standards, is limited to lower acceleration impacts. However, the more resilient 3rd generation Hybrid headform (HIII), originally designed for automotive crash testing, allows for higher acceleration impact testing. Mertz et. al. used the HIII headform to establish a series of injury probability curves, correlating peak linear acceleration and the HIC to the predicted risk of severe head injury (Mertz, 2003). With such injury tolerance data, the HIII is increasingly applied in biomechanical helmet and head impact investigations (Pellman, 2003; Viano, 2005; Pang, 2011; Scher, 2009).

The preliminary objective of this study was to validate the biomechanical response of the Hybrid III headform against that of the Magnesium headform currently used in helmet certification standards. This then allowed for the main objective of evaluating the biomechanical efficacy of bicycle helmets in reducing the risk of head injury in simulated bicycle fall or collision impacts. A custom built fixture allowing for attachment of both the Hybrid III and Magnesium headform to a monorail drop tower, facilitated comparison of both helmeted and unhelmeted drops. Linear acceleration and the head injury criterion were used to quantify the impact severity and Injury Assessment Reference Values were used to categorize the risk of injury.

METHODS

A linear drop monorail, similar to those specified in bicycle helmet certification standards, was used to simulate head impact using both the Hybrid III Headform (hereafter HIII, Humanetics ATD Manufacturing Inc., OH) and the Magnesium Headform (hereafter Mg, E960 size J half-headform, Cadex Inc., QC). The headform was connected to the monorail with a low friction, guide-follower system and custom-made ball arm to facilitate mounting of both the HIII and Mg. A flat, steel anvil mounted to the drop monorail acted as an impact surface for all drops. The headform was oriented such that the forehead impacted the anvil first.

In total, 24 drops were performed. To validate linear acceleration for the HIII, paired helmeted drops from prescribed heights were carried out with the HIII and Mg headforms. An additional eight drops using the unhelmeted HIII were then completed to compare with the helmeted HIII trials. The drop height was increased in 0.5m increments from 0.5m to a maximum of 3.0m for each test scenario. This height range brackets most of the current helmet certification standards (CSA D113.2-M89 (1.7m), CPSC (2.0 m), ASTM F1447 (2.0 m), EN1078 (1.5 m) Snell B95A (2.2 m)) and includes higher drop heights that may represent real-world bicycle accidents at considerable speed. Three drops were performed from a height of 2.0 m to allow for a limited investigation of repeatability through velocity and acceleration analysis. To account for frictional losses in the guide-follower system, 0.05m was added to the height for each drop. Impact velocity was calculated using high speed video and found to be within 5% of the expected velocity.

For each of the helmeted trials, a new bicycle helmet (CCM V15 Backtrail, micro shell and expanded polystyrene construction, Reebok-CCM Hockey, Montreal, QC, Canada) which conformed to the Consumer Product Safety Commission standards (CPSC, 1998) was fitted on the headform. Landmarks drawn on the headform ensured consistent positioning of the helmet for each trial. The chin retention strap was manually secured to eliminate free movement of the helmet. A ratcheting, micro-adjustable dial, which allow for customized circumferential fit and vertical position, was then tightened according to the manufacturer's specification.

Translational acceleration was measured during the impact with a single axis accelerometer (±2000 g range, Endevco model 7264C-2000, Meggitt Sensing Systems, San Juan Capistrano, CA, USA) mounted in the ball arm and placed in the headform near to its center of gravity. The accelerometer axis was aligned with the vertical axis of the drop rail.

An Analog Devices (Analog Devices Inc., Norwood MA) data acquisition system was used to collect the data. Acceleration was sampled at 39 kHz and the signal was processed with a low-pass filter at 1650 Hz in accordance with SAE J211-1. Each impact was recorded at 1000 frames/second using a Phantom V9 (Vision Research Inc., Wayne, NJ) high speed camera.

Given the two independent objectives of this investigation, helmeted HIII and Mg drops were first analyzed, and then a comparison of helmeted and unhelmeted HIII drops was made. The biomechanical metrics used to compare the drops included peak linear acceleration and the Head Injury Criterion (HIC) (equation 1). The HIC quantifies head impact severity by incorporating acceleration magnitude and time of acceleration exposure. For this analysis, (t_2-t_1) was taken to be a maximum of 15 ms.

$$HIC = \left\{ \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} (t_2 - t_1) \right\}_{max} \quad (1)$$

To assess head injury risk associated with each drop, an Injury Assessment Reference Value (IARV) probability curve was used. Mertz et. al. identifies a 5% risk of skull fracture at a linear head acceleration threshold of 180g as well as a 5% risk of an AIS ≥4 head injury for the adult population at a HIC_{15} threshold of 700 (Mertz, 2003).

RESULTS

Hybrid III Headform Validation

The repeatability analysis included three drops from 2.0 m for each of the HIII and Mg headforms. The maximum difference over the repeated drops was found to be 4.1% and 5.6% from the mean for the HIII and Mg headform's, respectively. The maximum difference observed for peak head acceleration was 3.3% and 2.6% while HIC varied by 5.0% and 4.7% of the respective headform mean values.

When comparing the two acceleration impulses over time for the HIII and Mg, a similar abrupt increase followed by a sharp decrease was observed. For both headforms a transient, non-uniform acceleration profile concludes the impact event, representing secondary interaction between the anvil and helmeted headform. The peak head accelerations, calculated HIC values and impulse duration were generally found to be comparable for the HIII and Mg.

The peak acceleration was, on average, observed to be 5% higher for the Mg headform impacts (ranging from exact agreement to a maximum 11% disagreement) however, no correlation was found between difference in peak acceleration and drop height. Both the HIII and Mg headforms exhibited accelerations exceeding the IARV of 180g for drops higher than 1.5 m. This IARV represents a 5% chance of skull fracture (Mertz, 2003).

Calculated maximum HIC values were also found to be comparable for the HIII and Mg headforms with the HIII being 6.5% lower on average. The mean interval found to maximize HIC for the HIII and Mg headforms was 0.0043 and 0.0046, respectively. The HIC IARV for a 5% risk of AIS ≥4 head injury (=700) was observed to be exceeded for all drops 1.5 m or higher. According to published data by Mertz. et. al., which defines HIC thresholds for a probability of sustaining an AIS 4+ brain injury, all drops performed from 2.0 m or lower correlated to less than 50% probability (Mertz, 2003). The Mg headform was found to be 6% more susceptible to injury than the HIII headform, averaging over all drop heights.

Physical damage was also observed at various levels with severity generally corresponding to drop height. In all cases, plastic deformation and cracking of the expanded polystyrene (EPS) was seen near the point of impact. For the 3.0 m drop height, the micro shell cracked and shattered.

Helmeted vs. Unhelmeted Hybrid III Headform

In the repeatability investigation from a drop height of 2.0 m, it was found that the maximum difference of head impact velocity over repeated drops was 4.1% and 3.0% from the mean velocity for the helmeted and unhelmeted HIII headform, respectively. Peak acceleration was observed to be within 1.5% and 3.0% of the mean for the 2.0 m drops while the HIC values was within 6.0% and 5.0% of the mean HIC.

In comparing the impulse events for each of the drop scenarios, an increase in acceleration followed by a decrease in acceleration was observed for both helmeted and unhelmeted drops. However, the magnitude of peak acceleration reached was significantly lower for the helmeted headform. In both cases, this initial rise and fall was then followed by a transient acceleration response, representative of the secondary collisions between the headform and the anvil. The duration of the primary impact impulse was larger in helmeted drops than in unhelmeted.

For all drop heights, peak accelerations for the helmeted headform were smaller than the unhelmeted scenario. An average reduction in peak acceleration by a factor of 4.2 was observed over all drops for the helmeted headform. When compared to the IARV for a 5% risk of skull fracture (180 g), the unhelmeted drops were found to exceed the threshold in every drop while the helmeted drops remained below the threshold for the 1.5 m drop and lower (Mertz, 2003). A linear relationship between drop height and peak acceleration was realized.

Calculated HIC values also exhibited similar trends to peak acceleration, with the bicycle helmet reducing the HIC for every drop height. The mean interval required to maximize HIC was found to be 0.001 ms for the unhelmeted drops and 0.005 ms for the helmeted drops. These findings are consistent with the relatively wider (in the time domain) acceleration impulse discussed above.

According to the AIS ≥4 head injury probability curve published by Mertz et. al., all unhelmeted drops above 0.5 m correspond to a 100% probability (Mertz, 2003). In contrast, helmeted drops only exceeded a 50% probability at the 2.5 m and 3.0 m drop heights, with neither reaching 100%. The helmeted drops from 2.0 m and below all corresponded to probabilities lower than 35%.

DISCUSSION

Using a monorail drop tower with a custom fabricated ball arm, we were able to compare the helmeted magnesium headform, used for many helmet certification standards, to the helmeted Hybrid III headform, for which several biomechanical metrics quantifying injury risk have been developed. We then conducted paired helmeted and unhelmeted drops with the HIII headform to investigate the efficacy of a typical bicycle helmet in reducing linear acceleration and in turn the risk of head injury. Linear acceleration and the Head Injury Criterion were used to characterize the impact event and compared to Injury Assessment Reference Values (Mertz, 2003), allowing for head injury severity to be estimated. This study was performed with the intention of supporting or refuting existing epidemiological studies, which strongly promote bicycle helmets.

In comparing the linear acceleration responses between the two headforms, a clear similarity is observed. Difference in peak linear acceleration and HIC was found to be an average of 5% and 8%, respectively. In general, the magnesium headform exhibited higher peak accelerations and HIC values. With such correlation, the magnesium and HIII headforms can be seen as equivalent tools in the biomechanical investigation of helmet efficacy in linear, frontal impact scenarios.

Investigation into the efficacy of conventional, off the shelf bicycle helmets also yielded encouraging results. Paired helmeted and unhelmeted drops with the HIII headform exhibited dramatic evidence to support bicycle helmets as a method to minimize head injury risk. An average reduction of peak linear acceleration by a factor of 4 and a HIC by a factor of 6.5 was realized in the drop testing. Mertz et. al. identify a linear acceleration of 180 g to correlate to a 5% risk of skull fracture for an adult and a HIC value of 1434 to correlate to a 50% risk of an AIS ≥4 head injury (Mertz, 2003). With the exception of HIC value for the 0.5 m drop height, unhelmeted drops exceeded these thresholds in every drop. In contrast, the helmeted HIII headform only reached these values in the 2 m and higher drop heights for peak acceleration and 2.5 m and higher drop heights for HIC.

A typical cycling accident has been reported to occur at an average impact velocity of 20km/hr (5.6 m/s) (Fahlstedt, 2012), which is approximately represented with the 1.5 m drop height investigated. At this drop height, the conventional bicycle helmet tested provided significant benefit in reducing head injury. Given the IARV's, a fall without a helmet from this height would almost certainly result in an AIS ≥4 injury (eg. penetrating skull injuries leading to brain injury, large contusions, severe hematomas, diffuse axonal injury etc.) while a helmet reduces this risk to 10%. The higher drop heights included in this study allow comparison to the current helmet certification standards as well as provide limited insight into the helmet performance in higher velocity impacts. Although the benefit of helmet use, like seat belts and all injury prevention devices, seems to be finite when evaluated at these higher velocities, a real world fall or collision impact velocity of 7.7 m/s (3 m drop height) is uncommon (Fahlstedt, 2012).

Epidemiological data has provided a direct window into bicycle injuries and the apparent efficacy of helmets in reducing this risk. This investigation supports and compliments the statistical data with biomechanical evidence. Many epidemiological papers have received disapproval with critics stating publication and time bias skewed the results in favor of helmet use. With such analysis being limited to post-accident data, it is difficult to discern how helmets have influenced injury outcomes. This study provides insight into the impact event with direct comparison of biomechanical metrics for a helmeted and unhelmeted biofidelic headform.

The results from this investigation are consistent with other biomechanical studies testing the safety benefit of bicycle helmets. In 2012, Mattei et. al. performed paired helmeted and unhelmeted linear drops with pediatric skulls. A maximum reduction of 87% percent in mean acceleration was observed from drop heights of 6 inches (0.15 m) and 9 inches (0.23 m) (Mattei, 2012). Benz et. al. conducted a study in which child and adult bicycle helmets were fitted on anthropomorphic test headforms and dropped linearly from heights of 1 m and 1.5 m. The group concluded that helmets decreased HIC by a factor of 3 (Benz, 1993). Another investigation with headforms dropped onto flat and convex surfaces from 1 m and 2 m showed that unhelmeted impact correlated with significantly increased injury risk (70-99% compared to <1% for helmeted drops; for 1 m drops) (Hodgson, 1990).

While present results are consistent with with the above findings, it is difficult to compare the results due to several key strengths of the present study that are lacking in previous work. Performing drops from a large range of heights allows for a broad spectrum of real world scenarios, from falling from seated on a stationary bicycle to falling off the bicycle at speed, to be investigated as well as comparisons to be made with current certification standards. A strength of this work is that the HIII headform also allows for injury risk to be quantified from established functions, enabling direct relevance to the cycling community.

Although the study was designed to simulate real world cycling accidents, the results are limited to the controlled nature of biomechanical laboratory testing. Velocity, anatomical impact location, environmental impact surface and external forces are all variables in actuality but were limited to specific scenarios for the purpose of our testing.

CONCLUSIONS

Bicycle helmets are effective in significantly reducing the peak linear acceleration and HIC values, both of which have been correlated to severe head and brain injury risk. For a drop height corresponding to the velocity of a typical cycling accident (1.5 m ≈ 5.6 m/s), wearing a helmet was found to decrease the risk of severe brain injury from 99.9% to 9.3%. A contemporary helmet can transform a head impact that would result in severe life-long disability into an impact with little potential for skull fracture or severe brain injury.

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